ENDOCARDIAL MAPPING CATHETER

now U.S. Pat. No. 6,240,307,

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Technical Field

The invention discloses the apparatus and technique for forming a three-dimensional electrical map of the interior of a heart chamber, and a related technique for forming a two-dimensional subsurface map at a particular location in the endocardial wall.

Background Art

It is common to measure the electrical potentials present on the interior surface of the heart as a part of an electrophysiologic study of a patient's heart. Typically such measurements are used to form a two-dimensional map of the electrical activity of the heart muscle. An electrophysiologist will use the map to locate centers of ectopic electrical activity occurring within the cardiac tissues. One traditional mapping technique involves a sequence of electrical measurements taken from mobile electrodes inserted into the heart chamber and placed in contact with the surface of the heart. An alternative mapping technique takes essentially simultaneous measurements from a floating electrode array to generate a two-dimensional map of electrical potentials.

The two-dimensional maps of the electrical potentials at the endocardial surface generated by these traditional processes suffer many defects. Traditional systems have been limited in resolution by the number of electrodes used. The number of electrodes dictated the number of points for which the electrical activity of the endocardial surface could be mapped. Therefore, progress in endocardial mapping has involved either the introduction of progressively more electrodes on the mapping catheter or improved flexibility for moving a small mapping probe with electrodes from place to place on the endocardial surface. Direct contact with electrically active tissue is required by most systems in the prior art in order to obtain well conditioned

electrical signals. An exception is a non-contact
approach with spot electrodes. These spot electrodes
spatially average the electrical signal through their
conical view of the blood media. This approach

5 therefore also produces one signal for each electrode.
The small number of signals from the endocardial wall
will result in the inability to accurately resolve the
location of ectopic tissue masses. In the prior art,
iso-potentials are interpolated and plotted on a

10 rectilinear map which can only crudely represent the
unfolded interior surface of the heart. Such two-
dimensional maps are generated by interpolation
processes which "fill in" contours based upon a limited
set of measurements. Such interpolated two-dimensional

15 maps have significant deficiencies. First, if a
localized ectopic focus is between two electrode views
such a map will at best show the ectopic focus
overlaying both electrodes and all points in between and
at worst will not see it at all. Second, the two

20 dimensional map, since it contains no chamber geometry
information, cannot indicate precisely where in the
three dimensional volume of the heat chamber an
electrical signal is located. The inability to
accurately characterize the size and location of ectopic

25 tissue frustrates the delivery of certain therapies such
as "ablation".

Summary Disclosure

In general the present invention provides a method
30 for producing a high-resolution, three-dimensional map
of electrical activity of the inside surface of a heart
chamber.

The invention uses a specialized catheter system to
obtain the information necessary to generate such a map.

35 In general the invention provides a system and
method which permits the location of catheter electrodes
to be visualized in the three-dimensional map.

The invention may also be used to provide a two-dimensional map of electrical potential at or below the myocardial tissue surface.

Additional features of the invention will appear

5 from the following description in which the illustrative embodiment is set forth in detail in conjunction with the accompanying drawings. It should be understood that many modifications to the invention, and in particular to the preferred embodiment illustrated in these

10 drawings, may be made without departing from the scope of the invention.

Figure 1 is a schematic view of the system.

Figure 2 is a view of the catheter assembly placed in an endocardial cavity.

15 Figure 3 is a schematic view of the catheter assembly.

Figure 4 is a view of the mapping catheter with the deformable lead body in the collapsed position.

Figure 5 is a view of the mapping catheter with the

20 deformable lead body in the expanded position.

Figure 6 is a view of the reference catheter.

Figure 7 is a schematic view representing the display of the three-dimensional map.

Figure 8 is a side view of an alternate reference

25 catheter.

Figure 9 is a side view of an alternate reference catheter.

Figure 10 is a perspective view of an alternate distal tip.

30 Figure 11 is a schematic view representing the display of the subsurface two-dimensional map.

Figure 12 is a schematic flow chart of the steps in the method.

35 Detailed Disclosure

In general, the system of the present invention is used for mapping the electrical activity of the interior

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surface of a heart chamber 80. The mapping catheter
assembly 14 includes a flexible lead body 72 connected
to a deformable distal lead body 74. The deformable
distal lead body 74 can be formed into a stable space
5   filling geometric shape after introduction into the
heart cavity 80. This deformable distal lead body 74
includes an electrode array 19 defining a number of
electrode sites. The mapping catheter assembly 14 also
includes a reference electrode preferably placed on a
10 reference catheter 16 which passes through a central
lumen 82 formed in the flexible lead body 72 and the
distal lead body 74. The reference catheter assembly 16
has a distal tip electrode assembly 24 which may be used
to probe the heart wall. This distal contact electrode
15 assembly 24 provides a surface electrical reference for
calibration. The physical length of the reference
catheter 16 taken with the position of the electrode
array 19 together provide a reference which may be used
to calibrate the electrode array 19. The reference
20 catheter 16 also stabilizes the position of the
electrode array 19 which is desirable.

These structural elements provide a mapping
catheter assembly which can be readily positioned within
the heart and used to acquire highly accurate
25 information concerning the electrical activity of the
heart from a first set of preferably non-contact
electrode sites and a second set of in-contact electrode
sites.

The mapping catheter assembly 14 is coupled to
30 interface apparatus 22 which contains a signal generator
32, and voltage acquisition apparatus 30. Preferably,
in use, the signal generator 32 is used to measure the
volumetric shape of the heart chamber through impedance
plethysmography. This signal generator is also used to
35 determine the position of the reference electrode within
the heart chamber. Other techniques for characterizing
the shape of the heart chamber may be substituted.

5

Next, the signals from all the electrode sites on the
electrode array 19 are presented to the voltage
acquisition apparatus 30 to derive a three-dimensional,
instantaneous high resolution map of the electrical
5   activity of the entire heart chamber volume. This map
is calibrated by the use of a surface electrode 24. The
calibration is both electrical and dimensional. Lastly
this three-dimensional map, along with the signal from
an intramural electrode 26 preferably at the tip of the
10   reference catheter 16, is used to compute a two-
dimensional map of the intramural electrical activity
within the heart wall. The two-dimensional map is a
slice of the heart wall and represents the subsurface
electrical activity in the heart wall itself.
15   Both of these "maps" can be followed over time
which is desirable. The true three-dimensional map also
avoids the problem of spatial averaging and generates an
instantaneous, high resolution map of the electrical
activity of the entire volume of the heart chamber and
20   the endocardial surface. This three-dimensional map is
an order of magnitude more accurate and precise than
previously obtained interpolation maps. The two-
dimensional map of the intramural slice is unavailable
using prior techniques.
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Hardware Description
     Figure 1 shows the mapping system 10 coupled to a
patient's heart 12. The mapping catheter assembly 14 is
inserted into a heart chamber and the reference
30   electrode 24 touches the endocardial surface 18.
     The preferred array catheter 20 carries at least
twenty-four individual electrode sites which are coupled
to the interface apparatus 22. The preferred reference
catheter 16 is a coaxial extension of the array catheter
35   20. This reference catheter 16 includes a surface
electrode site 24 and a subsurface electrode site 26
both of which are coupled to the interface apparatus 22.

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It should be understood that the electrode site 24 can
be located directly on the array catheter. The array
catheter 20 may be expanded into a known geometric
shape, preferably spherical. Resolution is enhanced by
5   the use of larger sized spherical shapes. A balloon 77
or the like should be incorporated under the electrode
array 19 to exclude blood from the interior of the
electrode array 19. The spherical shape and exclusion
of blood are not required for operability but they
10 materially reduce the complexity of the calculations
required to generate the map displays.

The reference electrode 24 and/or the
reference catheter 16 serves several purposes. First
they stabilize and maintain the array 19 at a known
15 distance from a reference point on the endocardial
surface 18 for calibration of the shape and volume
calculations. Secondly, the surface electrode 24 is
used to calibrate the electrical activity measurements
of the endocardial surface 18 provided by the electrode
20 array 19.

The interface apparatus 22 includes a switching
assembly 28 which is a multiplexor to sequentially
couple the various electrode sites to the voltage
acquisition apparatus 30, and the signal generator
25 apparatus 32. These devices are under the control of a
computer 34. The voltage acquisition apparatus 30 is
preferably a 12 bit A to D convertor. A signal
generator 32 is also supplied to generate low current
pulses for determining the volume and shape of the
30 endocardial chamber using impedance plethysmography, and
for determining the location of the reference catheter.

The computer 34 is preferably of the "workstation"
class to provide sufficient processing power to operate
in essentially real time. This computer operates under
35 the control of software set forth in the flow charts of
Figures 12A and 12B.

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Catheter Description

Figure 2 shows a portion of the mapping catheter
assembly 14 placed into a heart chamber 80. The mapping
catheter assembly 14 includes a reference catheter 16
5 and an array catheter 20. In Figure 2 the array catheter
20 has been expanded through the use of a stylet 92 to
place the electrode array 19 into a stable and
reproducible geometric shape. The reference catheter 16
has been passed through the lumen 82 of the array
10 catheter 20 to place a distal tip electrode assembly 24
into position against an endocardial surface. In use,
the reference catheter 16 provides a mechanical location
reference for the position of the electrode array 19,
and the tip electrode assembly 24 provides an electrical
15 potential reference at or in the heart wall for the
mapping process.

Although the structures of Figure 1 are preferred
there are several alternatives within the scope of the
invention. The principle objective of the preferred
20 form of the catheter system is to reliably place a known
collection of electrode sites away from the endocardial
surface, and one or more electrode sites into contact
with the endocardium. The array catheter is an
illustrative structure for placing at least some of the
25 electrode sites away from the endocardial surface. The
array catheter itself can be designed to mechanically
position one or more electrode sites on the endocardial
surface. The reference catheter is a preferred
structure for carrying one or more electrode sites and
30 may be used to place these electrode sites into direct
contact with the endocardial surface.

It should be understood that the reference catheter
could be replaced with a fixed extension of the array
catheter and used to push a segment of the array onto
35 the endocardial surface. In this alternate embodiment
the geometric shape of the spherical array maintains the
other electrodes out of contact with the endocardial

9

surface.

    Figure 3 shows the preferred construction of the
mapping catheter assembly 14 in exaggerated scale to
clarify details of construction. In general, the array
5   catheter 20 includes a flexible lead body 72 coupled to
a deformable lead body 74. The deformable lead body 74
is preferably a braid 75 of insulated wires, several of
which are shown as wire 93, wire 94, wire 95 and wire
96. An individual wire such as 93 may be traced in the
10 figure from the electrical connection 79 at the proximal
end 81 of the flexible lead body 72 through the flexible
lead body 72 to the distal braid ring 83 located on the
deformable lead body 74. At a predetermined location in
the deformable lead body 74 the insulation has been
15 selectively removed from this wire 93 to form a
representative electrode site 84. Each of the several
wires in the braid 75 may potentially be used to form an
electrode site. Preferably all of the typically twenty-
four to one-hundred-twenty-eight wires in the braid 75
20 are used to form electrode sites. Wires not used as
electrode sites provide mechanical support for the
electrode array 19. In general, the electrode sites
will be located equidistant from a center defined at the
center of the spherical array. Other geometrical shapes
25 are usable including ellipsoidal and the like.

    The proximal end 81 of the mapping catheter
assembly 14 has suitable electrical connection 79 for
the individual wires connected to the various electrode
sites. Similarly the proximal connector 79 can have a
30 suitable electrical connection for the distal tip
electrode assembly 24 of the reference catheter 16 or
the reference catheter 16 can use a separate connector.
The distance 90 between the electrode array 19 and the
distal tip assembly 24 electrode can preferentially be
35 varied by sliding the reference catheter through the
lumen 82, as shown by motion arrow 85. This distance 90
may be "read" at the proximal end 81 by noting the

relative position of the end of the lead body 72 and the
proximal end of the reference catheter 16.

Figure 4 is a view of the mapping catheter with the
deformable lead body 74 in the collapsed position.

5   Figure 5 shows that the wire stylet 92 is attached
to the distal braid ring 83 and positioned in the lumen
82. Traction applied to the distal braid ring 83 by
relative motion of the stylet 92 with respect to the
lead body 72 causes the braid 75 to change shape. In

10   general, traction causes the braid 75 to move from a
generally cylindrical form seen in Figure 4 to a
generally spherical form seen best in Figure 2 and
Figure 5.

The preferred technique is to provide a stylet 92

15   which can be used to pull the braid 75 which will deploy
the electrode array 19. However, other techniques may
be used as well including an optional balloon 77 shown
as in Figure 3, which could be inflated under the
electrode array 19 thereby causing the spherical

20   deployment of the array 19. Modification of the braid
75 can be used to control the final shape of the array
19. For example an asymmetrical braid pattern using
differing diameter wires within the braid can
preferentially alter the shape of the array. The most

25   important property of the geometric shape is that it
spaces the electrode sites relatively far apart and that
the shape be predictable with a high degree of accuracy.

Figure 6 shows a first embodiment of the reference
catheter 16 where the distal electrode assembly 24 is

30   blunt and may be used to make a surface measurement
against the endocardial surface. In this version of the
catheter assembly the wire 97 (Figure 2) communicates to
the distal tip electrode and this wire may be terminated
in the connector 79.

35   Figure 8 shows an alternate reference catheter 98
which is preferred if both surface and/or subsurface
measurements of the potential proximate the endocardial

surface are desired. This catheter 98 includes both a reference electrode 24 and an extendable intramural electrode body 100.

Figure 9 illustrates the preferred use of an
5 intramural electrode stylet 101 to retract the sharp intramural electrode body 100 into the reference catheter lead body 102. Motion of the intramural electrode body 100 into the lead body 102 is shown by arrow 103.

10 Figure 10 shows the location of the intramural electrode site 26 on the electrode body 100. It is desirable to use a relatively small electrode site to permit localization of the intramural electrical activity.

15 The array catheter 20 may be made by any of a variety of techniques. In one method of manufacture, the braid 75 of insulated wires 93,94,95,96 can be encapsulated into a plastic material to form the flexible lead body 72. This plastic material can be any
20 of various biocompatible compounds with polyurethane being preferred. The encapsulation material for the flexible lead body 72 is selected in part for its ability to be selectively removed to expose the insulated braid 75 to form the deformable lead body 74.
25 The use of a braid 75 rather than a spiral wrap, axial wrap, or other configuration inherently strengthens and supports the electrodes due to the interlocking nature of the braid. This interlocking braid 75 also insures that, as the electrode array 19 deploys, it does so with
30 predictable dimensional control. This braid 75 structure also supports the array catheter 20 and provides for the structural integrity of the array catheter 20 where the encapsulating material has been removed.

35 To form the deformable lead body 74 at the distal end of the array catheter 20, the encapsulating material can be removed by known techniques. In a preferred
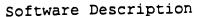
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embodiment this removal is accomplished by mechanical removal of the encapsulating material by grinding or the like. It is also possible to remove the material with a solvent. If the encapsulating material is polyurethane,

5   tetrahydrofuran or cyclohexanone can be used as a solvent. In some embodiments the encapsulating material is not removed from the extreme distal tip to provide enhanced mechanical integrity forming a distal braid ring 83.

10   With the insulated braid 75 exposed, to form the deformable lead body 74 the electrodes sites can be formed by removing the insulation over the conductor in selected areas. Known techniques would involve mechanical, thermal or chemical removal of the

15   insulation followed by identification of the appropriate conducting wire at the proximal connector 79. This method makes it difficult to have the orientation of the proximal conductors in a predictable repeatable manner. Color coding of the insulation to enable selection of

20   the conductor/electrode is possible but is also difficult when large numbers of electrodes are required. Therefore it is preferred to select and form the electrode array through the use of high voltage electricity. By applying high voltage electricity

25   (typically 1-3 KV) to the proximal end of the conductor and detecting this energy through the insulation it is possible to facilitate the creation of the electrode on a known conductor at a desired location. After localization, the electrode site can be created by

30   removing insulation using standard means or by applying a higher voltage (eg. 5 KV) to break through the insulation.

        Modifications can be made to this mapping catheter assembly without departing from the teachings of the

35   present invention. Accordingly the scope of the invention is only to be limited only by the accompanying claims.

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Software Description

The illustrative method may be partitioned into
nine steps as shown in Figure 12. The partitioning of
the step-wise sequence is done as an aid to explaining
the invention and other equivalent partitioning can be
5 readily substituted without departing from the scope of
the invention.

At step 41 the process begins. The illustrative
process assumes that the electrode array assumes a known
10 spherical shape within the heart chamber, and that there
are at least twenty-four electrodes on the electrode
array 19. This preferred method can be readily modified
to accommodate unknown and non-reproducible, non-
spherical shaped arrays. The location of each of these
15 electrode sites on the array surface is known from the
mechanical configuration of the displayed array. A
method of determining the location of the electrode
array 19 and the location of the heart chamber walls
(cardiac geometry) must be available. This geometry
20 measurement (options include ultrasound or impedance
plethysmography) is performed in step 41. If the
reference catheter 16 is extended to the chamber wall 18
then its length can be used to calibrate the geometry
measurements since the calculated distance can be
25 compared to the reference catheter length. The geometry
calculations are forced to converge on the known spacing
represented by the physical dimensions of the catheters.
In an alternative embodiment reference electrode 24 is
positioned on array catheter 20 and therefore its
30 position would be known.

In step 42 the signals from all the electrode sites
in the electrode array 19 are sampled by the A to D
converter in the voltage acquisition apparatus 30.
These measurements are stored in a digital file for
35 later use in following steps. At this point (step 43)
the known locations of all the electrodes on the
electrode array 19 and the measured potentials at each

/3

electrode are used to create the intermediate parameters
of the three-dimensional electrical activity map. This
step uses field theory calculations presented in greater
detail below. The components which are created in this
5   step (Φ_{lm}) are stored in a digital file for later use in
following steps.

At the next stage the question is asked whether the
reference catheter 16 is in a calibrating position. In
the calibrating position, the reference catheter 16
10 projects directly out of the array catheter 20
establishing a length from the electrode array 19 which
is a known distance from the wall 18 of the heart
chamber. This calibration position may be confirmed
using fluoroscopy. If the catheter is not in position
15 then the process moves to step 45, 46 or 47.

If the reference catheter 16 is in the calibrating
position then in step 44 the exact position of the
reference catheter 16 is determined using the distance
and orientation data from step 41. The available
20 information includes position in space of the reference
catheter 16 on the chamber wall 18 and the intermediate
electrical activity map parameters of the three-
dimensional map. Using these two sets of information
the expected electrical activity at the reference
25 catheter surface electrode site 24 is determined. The
actual potential at this site 24 is measured from the
reference catheter by the A to D converter in the
voltage acquisition apparatus 30. Finally, a scale
factor is adjusted which modifies the map calculations
30 to achieve calibrated results. This adjustment factor
is used in all subsequent calculations of electrical
activity.

At step 47 the system polls the user to display a
three-dimensional map. If such a map is desired then a
35 method of displaying the electrical activity is first
determined. Second an area, or volume is defined for
which the electrical activity is to be viewed. Third a

/14

14

level of resolution is defined for this view of the
electrical activity. Finally the electrical activity at
all of the points defined by the display option, volume
and resolution are computed using the field theory
5   calculations and the adjustment factor mentioned above.
These calculated values are then used to display the
data on computer 34.

Figure 7 is a representative display 71 of the
output of process 47. In the preferred presentation the
10 heart is displayed as a wire grid 36. The iso-potential
map for example is overlaid on the wire grid 36 and
several iso-potential lines such as iso-potential or
isochrone line 38 are shown on the drawing. Typically
the color of the wire grid 36 and the iso-potential or
15 isochrone lines will be different to aid interpretation.
The potentials may preferably be presented by a
continuously filled color-scale rather than iso-
potential or isochrone lines. The tightly closed iso-
potential or isochrone line 39 may arise from an ectopic
20 focus present at this location in the heart. In the
representative display 71 of process 47 the mapping
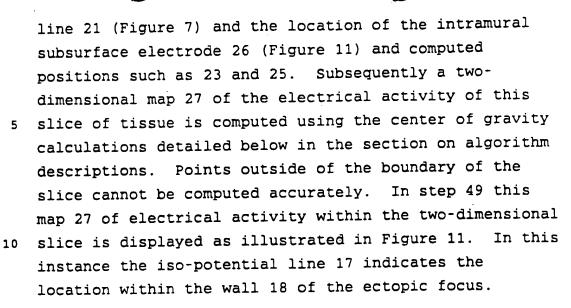catheter assembly will not be shown.

In step 45 a subthreshold pulse is supplied to the
surface electrode 24 of the reference catheter 16 by the
25 signal generator 32. In step 54 the voltages are
measured at all of the electrode sites on the electrode
array 19 by the voltage acquisition apparatus 30. One
problem in locating the position of the subthreshold
pulse is that other electrical activity may render it
30 difficult to detect. To counteract this problem step 55
starts by subtracting the electrical activity which was
just measured in step 44 from the measurements in step
54. The location of the tip of the reference catheter
16 (i.e. surface electrode 24), is found by first
35 performing the same field theory calculations of step 45
on this derived electrode data. Next, four positions in
space are defined which are positioned near the heart

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chamber walls. The potentials at these sites are
calculated using the three-dimensional electrical
activity map. These potentials are then used to
triangulate, and thus determine, the position of the
5   subthreshold pulse at the surface electrode 24 of the
reference catheter 16. If more accurate localization is
desired then four more points which are much closer to
the surface electrode 24 can be defined and the
triangulation can be performed again. This procedure
10   for locating the tip of the reference catheter 16 can be
performed whether the surface electrode 24 is touching
the surface or is located in the blood volume and is not
in contact with the endocardial surface.

      At step 48 the reference catheter's position in
15   space can be displayed by superimposing it on the map of
electrical activity created in step 47. An example of
such a display 71 is presented in Figure 7.

      When step 46 is reached the surface electrode 24 is
in a known position on the endocardial surface 18 of the
20   heart chamber which is proper for determining the
electrical activity of the tissue at that site. If the
intramural or subsurface extension 100 which
preferentially extends from the tip of the reference
catheter 102 is not inserted into the tissue then the
25   user of the system extends the subsurface electrode 26
into the wall 18. The potentials from the surface
electrode 24 and from the intramural subsurface 26
electrode are measured by voltage acquisition apparatus
30. Next a line 21 along the heart chamber wall which
30   has the surface electrode 24 at its center is defined by
the user of the system. The three-dimensional map
parameters from step 43 are then used to compute a
number of points along this line including the site of
the reference catheter surface electrode 24. These
35   calculations are adjusted to conform to the measured
value at the reference catheter surface electrode 24.
Next a slice of tissue is defined and bounded by this

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line 21 (Figure 7) and the location of the intramural
subsurface electrode 26 (Figure 11) and computed
positions such as 23 and 25. Subsequently a two-
dimensional map 27 of the electrical activity of this
5 slice of tissue is computed using the center of gravity
calculations detailed below in the section on algorithm
descriptions. Points outside of the boundary of the
slice cannot be computed accurately. In step 49 this
map 27 of electrical activity within the two-dimensional
10 slice is displayed as illustrated in Figure 11. In this
instance the iso-potential line 17 indicates the
location within the wall 18 of the ectopic focus.

Description of the Preferred Computing Algorithms
15   Two different algorithms are suitable for
implementing different stages of the present invention.
       The algorithm used to derive the map of the
electrical activity of the heart chamber employs
electrostatic volume-conductor field theory to derive a
20 high resolution map of the chamber volume. The second
algorithm is able to estimate intramural electrical
activity by interpolating between points on the
endocardial surface and an intramural measurement using
center of gravity calculations.
25   In use, the preliminary process steps identify the
position of the electrode array 19 consequently the
field theory algorithm can be initialized with both
contact and non-contact type data. This is one
difference from the traditional prior art techniques
30 which require either contact or non-contact for accurate
results, but cannot accommodate both. This also permits
the system to discern the difference between small
regions of electrical activity close to the electrode
array 19 from large regions of electrical activity
35 further away from the electrode array 19.
       In the first algorithm, from electrostatic
volume-conductor field theory it follows that all the

17

electrodes within the solid angle view of every locus of
electrical activity on the endocardial surface are
integrated together to reconstruct the electrical
activity at any given locus throughout the entire volume
5   and upon the endocardium. Thus as best shown in Figure
7 the signals from the electrode array 19 on the
catheter 20 produce a continuous map of the whole
endocardium. This is another difference between the
present method and the traditional prior art approach
10 which use the electrode with the lowest potential as the
indicator of cardiac abnormality. By using the complete
information in the algorithm, the resolution of the map
shown in Figure 7 is improved by at least a factor of
ten over prior methods. Other improvements include: the
15 ability to find the optimal global minimum instead of
sub-optimal local minima; the elimination of blind spots
between electrodes; the ability to detect abnormalities
caused by multiple ectopic foci; the ability to
distinguish between a localized focus of electrical
20 activity at the endocardial surface and a distributed
path of electrical activity in the more distant
myocardium; and the ability to detect other types of
electrical abnormalities including detection of ischemic
or infarcted tissue.
25   The algorithm for creating the 3D map of the
cardiac volume takes advantage of the fact that
myocardial electrical activity instantaneously creates
potential fields by electrotonic conduction. Since
action potentials propagate several orders of magnitude
30 slower than the speed of electrotonic conduction, the
potential field is quasi-static. Since there are no
significant charge sources in the blood volume,
Laplace's Equation for potential completely describes
the potential field in the blood volume:

18

$$\nabla^2\phi=0$$

LaPlace's equation can be solved numerically or
analytically. Such numerical techniques include
boundary element analysis and other interative
5   approaches comprised of estimating sums of nonlinear
coefficients.

Specific analytical approaches can be developed
based on the shape of the probe (i.e. spherical, prolate
10 spherical or cylindrical). From electrostatic field
theory, the general spherical harmonic series solution
for potential is:

$$\phi(r,\theta,\varphi) = \sum_{0}^{l=0} \sum_{m=-l}^{l} \{ A_l r^l + B_l r^{-(l-1)} \} \phi_{lm} Y_{lm}(\theta,\varphi)$$

In spherical harmonics, $Y_{lm}(\theta,\varphi)$ is the spherical
15 harmonic series made up of Legendre Polynomials. Φ_{lm} is
the lm^{th} component of potential and is defined as:

$$\phi_{lm} = \int V(\theta,\varphi) Y_{lm}(\theta,\varphi) \, d\Omega$$

where $V(\theta,\varphi)$ is the measured potential over the probe
radius R and $d\Omega$ is the differential solid angle and, in
spherical coordinates, is defined as:

$$d\Omega = \sin\theta \, d\theta \, d\varphi$$

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/9

During the first step in the algorithmic
determination of the 3D map of the electrical activity
each Φlm component is determined by integrating the
potential at a given point with the spherical harmonic
5   at that point with respect to the solid angle element
subtended from the origin to that point. This is an
important aspect of the 3D map; its accuracy in creating
the 3D map is increased with increased numbers of
electrodes in the array and with increased size of the
10   spherical array. In practice it is necessary to compute
the Φ_{lm} components with the subscript 1 set to 4 or
greater. These Φ_{lm} components are stored in an 1 by m
array for later determination of potentials anywhere in
the volume within the endocardial walls.

15   The bracketed expression of equation 1 (in terms of
A_1, B_1, and r) simply contains the extrapolation
coefficients that weight the measured probe components
to obtain the potential components anywhere in the
cavity. Once again, the weighted components are summed
20   to obtain the actual potentials. Given that the
potential is known on the probe boundary, and given that
the probe boundary is non-conductive, we can determine
the coefficients A_1 and B_1, yielding the following final
solution for potential at any point within the
25   boundaries of the cavity, using a spherical probe of
radius R:

$$\phi(r,\theta,\varphi) = \sum_{l=0}^{\infty}\sum_{m=-l}^{l}\left[\left(\frac{l+1}{2l+1}\right)\left(\frac{r}{R}\right)^{l}+\left(\frac{l}{2l+1}\right)\left(\frac{r}{R}\right)^{-l-1}\right]\phi_{lm}Y_{lm}(\theta,\varphi)$$

On exemplary method for evaluating the integral for
Φ_{lm} is the technique of Filon integration with an
30   estimating sum, discretized by p latitudinal rows and q
longitudinal columns of electrodes on the spherical
probe.

T0210

$$\phi_{lm} \geq \frac{4\pi}{pq} \sum_{i=1}^{p} \sum_{j=1}^{q} V(\theta_i, \varphi_j) Y_{lm}(\theta_i, \varphi_j)$$

Note that p times q equals the total number of
electrodes on the spherical probe array. The angle θ
ranges from zero to π radians and φ ranges from zero to
2π radians.

5   At this point the determination of the geometry of
the endocardial walls enters into the algorithm. The
potential of each point on the endocardial wall can now
be computed by defining them as r, θ, and φ. During the
activation sequence the graphical representation of the
10   electrical activity on the endocardial surface can be
slowed down by 30 to 40 times to present a picture of
the ventricular cavity within a time frame useful for
human viewing.

          A geometric description of the heart structure is
15   required in order for the algorithm to account for the
inherent effect of spatial averaging within the medium
(blood). Spatial averaging is a function of both the
conductive nature of the medium as well as the physical
dimensions of the medium.

20   Given the above computed three-dimensional
endocardial potential map, the intramural activation map
of Figure 11 is estimated by interpolating between the
accurately computed endocardial potentials at locations
23 and 25 (Figure 7), and actual recorded endocardial
25   value at the surface electrode 24 and an actual recorded
intramural value at the subsurface electrode 26 site.
This first-order estimation of the myocardial activation
map assumes that the medium is homogenous and that the
medium contains no charge sources. This myocardial
30   activation estimation is limited by the fact that the
myocardial medium is not homogeneous and that there are
charge sources contained within the myocardial medium.
If more than one intramural point was sampled the

21

underlying map of intramural electrical activity could
be improved by interpolating between the endocardial
surface values and all the sample intramural values.
The center-of-gravity calculations can be summarized by
5 the equation:

$$V(\overline{r_x}) = \frac{\sum\limits_{i=1}^{n} V_i \left(|\overline{r_x} - \overline{r_i}|^{-k}\right)}{\sum\limits_{i=1}^{n} |\overline{r_x} - \overline{r_i}|^{-k}}$$

where, $V(_x)$ represents the potential at any desired point
defined by the three-dimensional vector $_x$ and, V_i
represents each of n known potentials at a point defined
10 by the three-dimensional vector $_i$ and, k is an exponent
that matches the physical behavior of the tissue medium.

From the foregoing description, it will be apparent
that the method for determining a continuous map of the
electrical activity of the endocardial surface of the
15 present invention has a number of advantages, some of
which have been described above and others of which are
inherent in the invention. Also modifications can be
made to the mapping probe without departing from the
teachings of the present invention. Accordingly the
20 scope of the invention is only to be limited as
necessitated by the accompanying claims.